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Gongchen Sun, Zehao Pan, Satyajyoti Senapati, and Hsueh-Chia Chang

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Concentration Gradient Stabilization with Segregated Counter- and Co-Ion Paths: A Quasi-Stationary Depletion Front for Robust Molecular Isolation/Concentration

Gongchen Sun, Zehao Pan, Satyajyoti Senapati, and Hsueh-Chia Chang*
Department of Chemical and Biomolecular Engineering,
University of Notre Dame, Notre Dame, IN 46556, USA
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We study the spatiotemporal dynamics of a microfluidic system with a non-selective microfluidic channel gated by an ion-selective membrane which separates the ion flux paths of cations and anions. To preserve electro-neutrality, the ionic concentration in the system is shown to converge to a specific inhomogeneous distribution with robust constant current fluxes. A circuit scaling theory that collapses measured asymptotic currents verifies that this is a generic and robust mechanism insensitive to channel geometry, ion-selectivity and electrolyte ionic strength. This first temporally stationary but spatially inhomogeneous depletion front can be used for modulating ionic current and for isotachophoretic isolation of low-mobility molecules and exosomes on small diagnostic chips for various medical applications that require robust high-throughput and integrated platforms.

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I. INTRODUCTION

The spatiotemporal dynamics of ions and currents in a microfluidic system with adjoining ion-selective media has long been suspected to be driven by the electro-neutrality constraint. Without electric field or current penetration through the channel walls or the ion-selective media, the field lines and the current lines within a system are identical [1]. Both cations and anions share the same flux lines, though in opposite directions. The net tangential cation and anion fluxes remain the same at every position along the field line and cancel each other. This flux invariance condition can be specified by the same local tangential electric field along the field/current lines with a spatially homogeneous electrolyte concentration field [1]. Thus electro-neutrality can be preserved without spatial inhomogeneity. However, if an ion-selective membrane separates the flux lines of different ions, or if there is field leakage through the wall [2], the field lines separate from the current lines, which then segregate into two distinct cation and anion flux lines. Under this condition, flux invariance along different field lines cannot be achieved for both ions if the system remains spatially homogeneous. Hence, electro-neutrality necessarily requires spatial inhomogeneity for such systems with ion-selective media. In his seminal monograph, Rubinstein [3] shows that a diffusion layer with a non-uniform ionic strength appears on one side of an ion-selective membrane under a voltage bias to preserve electro-neutrality. A recent analysis by Yariv and Almog [4] (that generalizes the analysis of Ben and Chang [5] of the diffusive current towards the ion-selective membrane) attributes a memory effect in the ion concentration polarization dynamics to electro-neutrality.

On the other hand, spatially inhomogeneous ionic concentration polarization is used in many existing biomedical applications and has been suggested for future ones. Examples in current biotechnology include isoelectric focusing, isotachophoresis and field-flow fractionation. Ion current rectification across a conic nanopore was recently attributed to a gradient of the mobile ion concentration along the pore [6]. This phenomenon could lead to artificial biomimetic ion channels for control and quantification of specific ions or molecules [7]. Indeed, microfluidic chips with integrated perm-selective membrane modules have been designed to produce on-chip concentration gradients for the control and analysis of macromolecules, such as the transport, concentration and identification of protein and nucleic acid biomarkers for autonomous liquid biopsy platforms in precision medicine [8]. Low concentrations and mobility of these biomarkers require an on-chip analytical system that can isolate the molecules in a short channel. Recent biosensor work [9] shows that the ionic strength jump across a depletion front generated by an ion-selective membrane can be used for isotachophoretic isolation of molecules, exosomes and nanoparticles. A common problem in all these on-chip depletion front technologies is that, without the presence of large reservoirs, the on-chip inhomogeneous concentration profile is highly dynamic and unstable. A stationary ionic strength inhomogeneity (a stationary depletion front) that does not require pairing of electrolytes with different salts or pH can produce robust isotachophoretic and isoelectric purification and isolation platforms for low-mobility analytes in a short channel/biochip.

This concentration depletion front is governed by the diffusion equation and can propagate to infinity (or the end of the channel) without reaching a steady state [10]. Current work focuses on using a counter bulk flow, such as an electro-osmotic flow or convective mixing [9, 11–15], to balance the advance of the depletion front. However, the counter bulk flow will likely weaken the necessary inhomogeneity and cause dispersion of the ana-

* Corresponding author.
hchang@nd.edu

lyte. Meanwhile, most counter bulk flow methods such as using an electro-osmotic flow are only effective in shallow microchannels (a few μm), and cannot allow high-throughput processing of multiple samples in many medical applications.

In this paper, we report a particular coupled gating geometry between two different ionic paths in a microfluidic chip that can drive the depletion front from an ion-selective membrane to a quasi-stationary position, by only using the eletro-neutrality constraint and without introducing a counter bulk flow. The spatial inhomogeneity of the ion concentration in the microchannel stays stable when the trans-membrane potential drop and current vanish. This strategy of establishing a quasi-stationary depletion front can potentially be used to design on-chip isotachophoretic and isoelectric sample pretreatment modules in integrated medical diagnostic platforms.

II. FABRICATION AND EXPERIMENTAL METHODS

Our system with two different ionic paths consists of a non-selective microfluidic channel that allows both cation and anion electro-migration fluxes and an anion exchange membrane (AEM) that only allows counter-ion

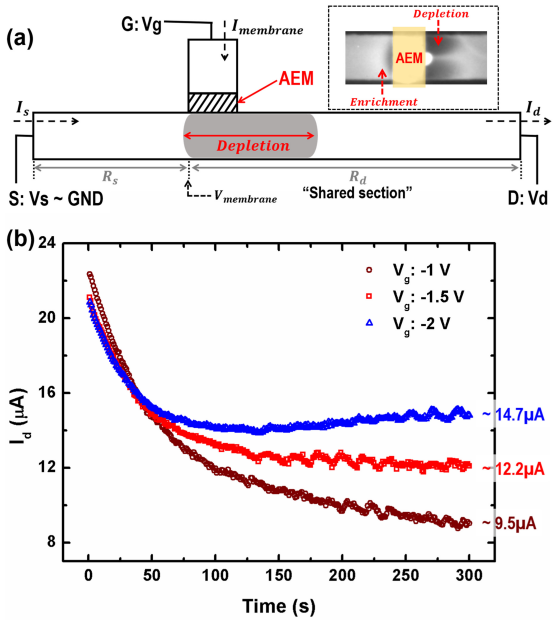

FIG. 1. (a) Schematic of the hybrid ionic path device. The inset shows fluorescent imaging of simultaneous ion depletion and enrichment on the same side of the AEM membrane. The image was taken from the bottom of the cross-channel, with $V_d = -15V$ and $V_g = -3V$. (b) Temporal evolution of cross-channel current I_d with different V_g ($-1V$, $-1.5V$, $-2V$), approaching a different steady-state current for different V_g. $V_d = -20V$ in all three experiments.

(anion) fluxes. These two ionic paths, one ion-selective and one non-selective, intersect perpendicularly, as illustrated in Fig. 1(a) and the fluxes are driven by three electrodes: a gating electrode G outside the membrane and a drain-source (D-S) pair in the cross-channel. These two ionic paths share a common section within the microfluidic channel (Fig. 1(a)). As a result of this coupling, the charge flux through the selective AEM can be arrested to preserve electro-neutrality if the voltage in the shared section $V_{membrane}$ approaches the gating voltage V_g. Once the charge flux through the membrane is arrested, the depletion front stops propagating at a quasi-stationary position.

In order to prove that our quasi-stationary depletion front is robust in different microfluidic systems, we fabricate two kinds of hybrid ionic path device with different geometrical parameters: an AEM based polycarbonate (PC) microfluidic chip and an AEM based polydimethylsiloxane (PDMS) microfluidic chip.

The AEM based PC microfluidic chip consists of three layers of PC sheets, and microchannels are structured by cutting the middle PC sheet on a plotter (Graphtec Cutting Pro FC7000MK2-60). The openings of microchannels for fluidic connections and membrane attachment are cut on the top PC sheet. These three structured PC sheets are then aligned and thermally bonded together at 170°C for 30 minutes. Channels with two different dimensions are fabricated: 2mm width × 28mm length × 250μm height and 1mm width × 28mm length × 250μm height to study the arrest of depletion front in different non-selective ionic paths discussed later in this paper. Cut pipette tips with a filter paper at the bottom as buffer reservoirs for electrical connection and Tygon tubings as fluidic inlets and outlets are fixed by a UV curable glue (Acrifix 192) onto their designated places on the top PC sheet. A layer of 1% agarose gel is placed on the bottom of each cut pipette tip reservoir to prevent possible bubble entry and to suppress electro-osmotic flows during electrical measurements. A 1mm × 10mm AEM strip (Mega a.s., Czech Republic) is cut and embedded into the membrane opening on the top PC sheet to cover the microfluidic channel at the designated position. The last step to complete the device is to fix and seal the membrane strip with the UV curable glue, making the membrane closely attached on top of the microfluidic channel and connecting two separated microchannels: one is used as the non-selective ionic path shown in Fig. 1(a), and the other one to apply gating voltage on the other side of the membrane. The AEM based PC microfluidic chips are used in experiments shown in Fig. 4.

The AEM based PDMS microfluidic chip is fabricated using the PDMS casting method against a glass/tape/membrane master of fluidic structures. A layer of double-sided Kapton tape with 100μm in thickness is cut with the plotter and then transferred onto a precleaned glass substrate as the channel mold. The mold defines a channel with 2mm width × 28mm length × 100μm height in dimension as the non-selective ionic

path. A 1mm × 10mm AEM strip is fixed onto the double-sided tape mold at the designated place. Similarly, the membrane strip connects two channels: one channel as the non-selective ionic path and the other one for gating voltage application. Silicone tubings for both fluidic inlets/outlets and electrical connections are positioned onto the double-sided tape mold to complete the casting master. PDMS prepolymer (Sylgard 184) is prepared by mixing base and curing agent in a weight ratio of 10:1 and is degassed in vacuum for 15 minutes. The PDMS prepolymer is then poured onto the completed master and degassed again. The whole structure is cured in an oven at 70°C or 50 minutes. The cured PDMS cast is peeled off and bonded onto a precleaned glass slide using corona discharge followed by thermal bonding at 70°C for 2 hours to complete the device. Both devices are filled with 10mM potassium chloride (KCL) solution for 48 hours to let the AEM swell properly prior to use. The AEM based PDMS microfluidic chips are used in experiments shown in Fig. 1 to Fig. 4.

To understand the dynamics of establishing the stationary depletion front, we perform a two-channel chrono-amperometry experiment on both microfluidic systems with real-time fluorescence imaging. A 10mM KCL solution is used as the electrolyte in all electrical characterization. External voltages are applied through platinum wires on the G, D and S ends (Fig. 1(a)) of the microfluidic system simultaneously. In each measurement, trans-membrane current $I_{membrane}$ measured from the G end, and cross-channel current I_d measured from the D end are recorded for over 300s by a Keithley 2636A Dual-Channel System SourceMeter Instrument. The gated and normalized ionic resistance of the cross-channel is subsequently calculated from the steady-state I_d. To visualize the ion concentration polarization region, we use 100μM cationic Rhodamine 6G dye dissolved in 10mM KCL solution as a fluorescence indicator without changing the conductivity of the buffer. The visualization process is performed on a customized dark room platform equipped with a Dark Reader Transilluminator (ClareChemical) to excite the fluorescent dye from the top of the microfluidic chip. Fluorescence images are taken from the bottom of the microfluidic chip by a QImaging Retiga 2000R Fast 1394 camera synchronized with the Keithley SourceMeter Instrument through a custom MATLAB code.

III. RESULTS AND DISCUSSION

We assign the S end of the cross-channel as the electric potential reference point (GND) and set a constant negative voltage V_d at the D end of the cross-channel to electrophoretically drive ion fluxes through the non-selective ionic path. By properly applying a negative potential V_g (between GND and V_d) at the G end of the anion selective path, an ion depletion zone and an ion enrichment zone can be generated simultaneously in the cross-channel [16]. The dynamics towards an electro-neutral state, with vanishing trans-membrane current (Fig. 2(c)), involve equilibration of the voltage $V_{membrane}$ that is at the junction of the two ionic paths (the shared section in Fig. 1(a)) towards V_g. The adjustment of $V_{membrane}$ during the equilibration results from propagation of the depletion front towards a specific location at the D end of the cross-channel. Unlike a homogeneous electrolyte which usually produces a linear electric potential profile along the channel, this depletion amplifies the ionic resistance and redistributes the electric potential along the cross-channel, until $V_{membrane} = V_g$.

In the actual device, due to the finite width of the AEM, the local value of $V_{membrane}$ varies slightly along the interface between the membrane and the cross-channel when the depletion front approaches the specific location. Although the net trans-membrane current across the membrane vanishes in the normal direction, such tangential potential difference of $V_{membrane}$ results in a tangential cross-membrane anion flux which enters the AEM from the left side of the cross-channel and exits to the right side. The established depletion zone to the right (D side) of the membrane can hence be maintained. However, this tangential cross-membrane anion flux is much smaller than the original trans-membrane flux, with far smaller space charge creation and anion depletion rates. As seen in the fluorescence image inset of Fig. 1(a), simultaneous ion enrichment to the left and depletion to the right occur on the channel side of the gating membrane, supporting the development of a tangential cross-membrane anion flux entering and exiting the finite-width membrane without penetration [17]. This tangential anion flux across the membrane sustains the depletion front position but not its original front speed.

Fig. 1(b) shows the cross-channel current evolution with V_d set at $-20V$ but with different V_g ranging from $-1V$ to $-2V$. I_d approaches different steady-state values for different V_g with slight variation at the end of all three experiments, indicating that the depletion front stabilizes and the cross-channel ionic resistance approaches specific values. The slight variation in current is due to a weak vortex instability developed on the depletion side of the membrane. However, by designing the cross-channel wall parallel to the AEM with a small separation (100μm) [10], we are able to suppress the vortex instability so that it does not arrest the depletion growth and disturb the stabilization of the cross-channel current.

For an unbounded symmetric electrolyte without two segregated current paths, the depletion front from an ion-selective membrane advances rapidly to infinity with a universal self-similar diffusive scaling $\sqrt{Dt}$ [18]. However, the tracking of the instantaneous length of the ion depletion region L_{dep} (by low intensity of the cationic fluorescence indicator) in our system with two different ionic paths ($V_g = -2.5V$ and $V_d = -15V$) indicates that initial diffusive dynamics develop into a slow non-diffusive one (Fig. 2(a)) at the same time when I_d approaches steady state. The fact that the initial depletion

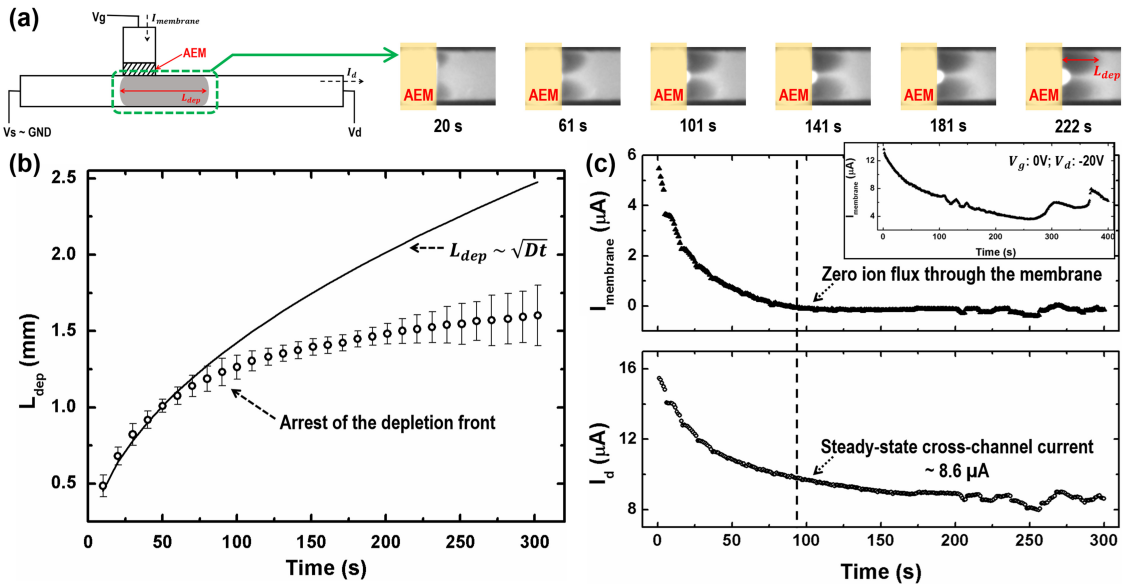

FIG. 2. (a) Fluorescence images of the depletion front at the indicated times. (b) Time evolution of the instantaneous depletion region length. Also depicted is the $\sqrt{Dt}$ scaling as a continuous line. A clear break from the self-similar scaling of the depletion length growth is observed after $90s$. (c) The charge-generating counter-ion flux through the AEM ($I_{membrane}$) decreases to zero after $90s$. Correspondingly, the cross-channel current I_d approaches a steady-state value beyond that time. $V_d = -15V$ and $V_g = -2.5V$ in the experiment. Inset shows the trans-membrane current $I_{membrane}$ becomes unstable at $V_g = 0V$ and $V_d = -20V$. $I_{membrane}$ cannot vanish to maintain system electro-neutrality since the voltage in the cross-channel across the membrane cannot reach the zero gating voltage.

front evolution follows the self-similar diffusive scaling indicates that the hydrodynamic vortex instabilities and convections are suppressed. However, the deviation of depletion growth from the diffusive scaling after $90s$ (Fig. 2(b)) suggests that the depletion front ceases to advance rapidly and approaches a stationary position to suppress space charge creation. The cationic fluorescent dyes can only migrate linearly with a limited electrophoretic mobility, when they enter the homogeneous low electric field region beyond the quasi-stationary depletion front after $t = 90s$. The arrested concentration polarization region is about 20mm away from the D end of the channel, far from the end boundary. The movie which shows the real-time growth and arrest of the depletion front can be found in Supplemental Materials [19]. Due to the negative surface charges of the PDMS microchannel [20], electro-osmotic flow may be generated at the cross-channel during the growth of the depletion front. In order to suppress net electro-osmotic convection, we seal the end-tubings for electric connections at both the S end and the D end of the cross-channel by glue [10]. The closed microchannel encapsulates the electrolyte and hence suppresses net electro-osmotic convections or amplified electrokinetic flows [21]. A pair of vortices due to a weak backflow from the D end can still be observed in Fig. 2(a) and causes small fluctuations in the cross-channel current (Fig. 2(c)). We can further eliminate the electro-osmotic instability by chemically modifying the PDMS microchannels to remove surface charges [22], or by filling the microchannel with hydrogels to block the flow [9].

The currents through these two ionic paths during the process of the depletion region evolution are measured simultaneously, shown in Fig. 2(c). As a result of the negative feedback from the depletion growth, $V_{membrane}$ approaching V_g renders the current flowing through the AEM anion selective path, $I_{membrane}$, to cease to zero after $90s$, which corresponds to vanishing of the fast charge-generating counter-ion flux to preserve electro-neutrality. Subsequently, the current through the non-selective ionic path approaches a steady-state value (8.6μA in this case), as the ionic resistance of the cross-channel approaches a specific value determined by the ion concentration pattern. There also exists zero or positive gating voltages that can never be reached by the voltage field in the cross-channel. Shown in the inset of Fig. 2(c), the trans-membrane current $I_{membrane}$ does not vanish when the gating voltage V_g is set to $0V$. Hence the electro-neutrality in the system can never be met, resulting in the space charge generation that can drive charging/discharging events, which are observed in the unstable $I_{membrane}$ after $250s$ in Fig. 2(c) (inset).

A series of experimental data correlating the steady-state ionic resistance of the cross-channel with the quasi-stationary depletion length are shown in Fig. 3 for constant V_d ($-15V$) but varying V_g from $-0.5V$ to $-3V$. The depletion length L_{dep} is extracted from the fluorescence image when the cross-channel current first reaches

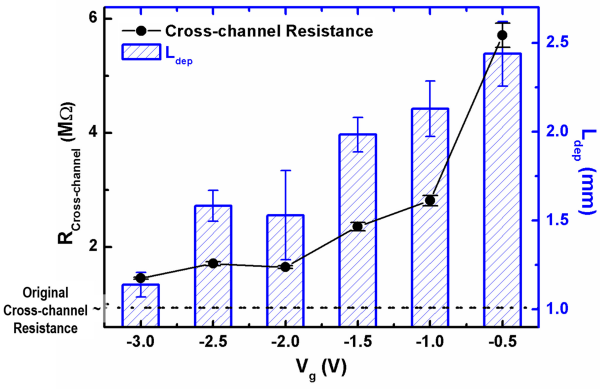

FIG. 3. The ionic resistance of the cross-channel shows strong correlation with the depletion region length and the gating voltage V_g. $V_d = -15V$ in all experiments.

the steady-state value. The depletion amplified ionic resistance $R_{Cross-channel}$ is compared with the original ionic resistance of the cross-channel without any gating effect which is calculated based on the channel geometry and the initial homogeneous electrolyte condition (Fig. 3), showing the increase of the ionic resistance due to the low mobile ion concentration within the depletion region. The consistent trend between $R_{Cross-channel}$ and L_{dep} versus V_g clearly confirms that the ionic resistance of the non-selective ionic path is modulated by the gating potential through the specific selection of the depletion length.

Equilibration of $V_{membrane}$ and V_g at steady state allows us to develop a simple scaling theory for the cross-channel resistance. Through ionic current balance $\frac{V_s - V_g}{R_s} = \frac{V_g - V_d}{R_d}$, and given that the cross-channel $R_{Cross-channel} = R_s + R_d$, we then have $R_{Cross-channel} = \frac{V_d}{V_g} R_s$. We define the length ratio between the distance from the S end to the AEM and the entire cross-channel length as f, so that the ratio between R_s and the original ionic resistance of the cross-channel is also f. Here we neglect the enrichment effect on the left side of the membrane, since under our voltage relationship we expect that the enrichment does not affect the ionic resistance as strongly as the depletion in the shared section. We can then normalize the $R_{Cross-channel}$ by the original ionic resistance of the cross-channel without any gating. So the normalized depletion amplified ionic resistance $\overline{R_{Cross-channel}} = f\frac{V_d}{V_g}$. For a given channel design, the amplification of the ionic resistance of the non-selective cross-channel at steady state depends only on the ratio of the applied voltages and the relative position of the ion-selective membrane. The normalized ionic resistance is not influenced by the channel dimension, electrolyte ionic strength and the type/strength of membrane selectivity, which makes this system versatile for various high-throughput medical applications that operate with complex buffers, large microchannels (up to a few hundred μm) and with different ion-selective membranes.

To verify this scaling model, we test seven groups of two-channel chrono-amperometry characterization against four different chips, each with different channel sizes, membrane positions and various potential combinations (Fig. 4). The experimental details can be found in Supplemental Materials [19]. The normalized ionic resistance of the cross-channel is extracted from different chips with various V_d and V_g combinations. After scaling, the large scattered data of these seven groups collapse to the curve predicted by the circuit model, as shown in Fig. 4(b).

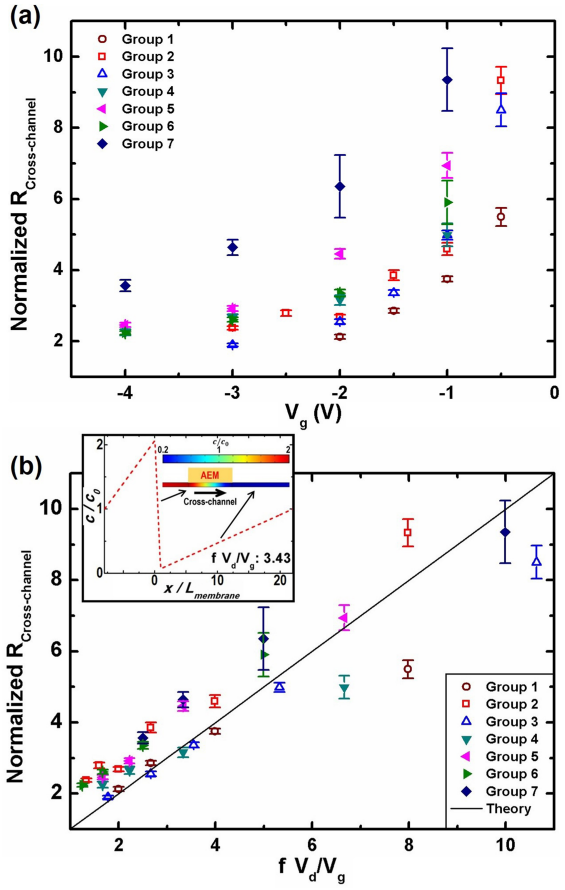

FIG. 4. (a) The normalized ionic resistance of the cross-channel increases with a higher gating voltage which results in a longer depletion region, regardless of the channel geometry. (b) Collapsed normalized ionic resistance data by the scaling theory with respect to a normalization factor that contains the gating potential V_g, draining potential V_d and the membrane position factor f. The parameter f is defined as the length ratio between the distance from the left end of the cross-channel to the membrane and the entire cross-channel length. Inset shows simultaneous enrichment and depletion at the vicinity of the membrane in the cross-channel by simulation, as is consistent with the fluorescent image of Fig. 1(a). Group 1-7 correspond to four different chip designs with different experimental set-ups. Detailed experimental information is provided in Supplemental Materials [19].

The inset of Fig. 4(b) illustrates the simulated ion concentration profile close to the AEM in the cross-channel when $f\frac{V_d}{V_g} = 3.43$, by using COMSOL Multiphysics v3.5a for the relevant Nernst-Planck equations for symmetric electrolytes. We use the model of Yossifon and Chang [11] for the finite-thickness ion-selective membrane. The simulated trans-membrane current at steady-state is near zero (∼38.6nA, which is two orders of magnitude less than the cross-channel current) and the co-existence of an ion enrichment region on the left of the membrane and ion depletion region on the right of the membrane can be seen in the simulated concentration profile shown in the inset of Fig. 4(b). The simulated image matches the experimental one shown by the fluorescence image in the inset of Fig. 1(a), validating the proposed mechanism that the rapidly advanced depletion front develops into a slowly-evolving one around a quasi-stationary position to preserve electro-neutrality.

IV. CONCLUSIONS

In summary, this paper presents a strategy to establish a quasi-stable but non-uniform ion concentration distribution in a microfluidic system. By gating a non-selective microfluidic channel with an ion-selective membrane, an ion depletion front is generated from the ion-selective membrane and subsequently arrested at a quasi-stationary position in the microfluidic channel due to the global electro-neutrality constraint. We can precisely control the position of the quasi-stationary depletion front and hence select the steady-state ionic resistance of the non-selective microchannel by only assigning an appropriate combination between the external gating voltage V_g and draining voltage V_d. Because counter bulk flows commonly used to stop the advance of the depletion front are not needed in this approach, we are able to simplify the design of the microfluidic system and to minimize analyte dispersion. A circuit scaling theory based on ionic current balance and the electro-neutrality constraint has been developed to analyze the control of the depletion-amplified ionic resistance through external voltages. The proposed mechanism to realize temporally slowly-evolving but spatially inhomogeneous ion concentration distribution is robust and insensitive to channel geometry, ion-selectivity and electrolyte ionic strength of the microfluidic system. Hence it can be adapted into a wide range of applications with complex buffers and large microchannels. In particular, we anticipate that this strategy can lead to a new on-chip isotachophoretic sample pretreatment device for purification, separation and concentration of low-mobility biomarkers like nucleic acids and exosomes. Such a device should prove invaluable in future high-throughput and integrated liquid biopsy platforms.

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